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ST T15 T20 T3
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2S12 2S19 2S1C 2S1D
2S1E 2S23 2S27 2S2B
2S2C 2S2D 2S2P 2S4D
2S5 2S7C 2SU1 5H2L 9N3
EBX
H4F D12X D18X D27M
D30B D30K D56X D61
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GB 1351026
GB 1154825
GB 1095162
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(71) Applicants
The Regents of the University of California, 2200 University Avenue, Berkeley, California, United States of America
(72) Inventors
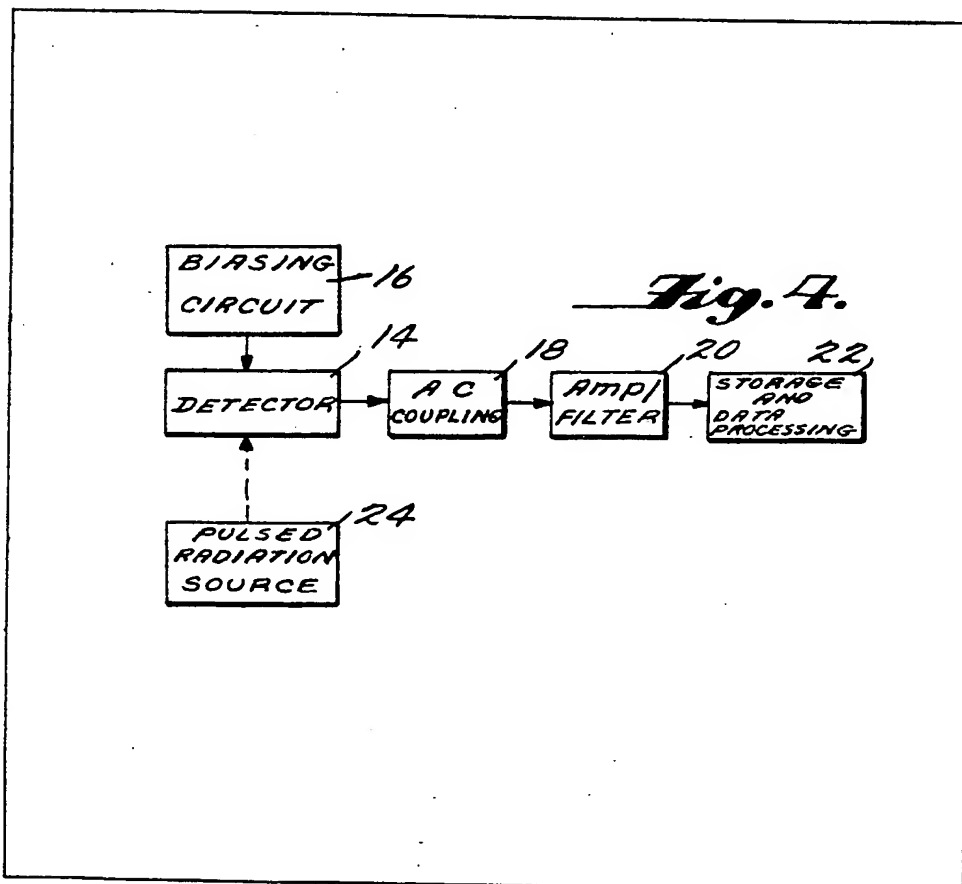**Leon Kaufman, Kenneth Edward Hosier**
(74) Agents
Stevens, Hewlett & Perkins

(54) **Detection in x-ray tomographic scanner**

(57) An electrical charge amplifier 20 including a filter circuit is AC coupled to a semiconductor (CdTe) detector 14 and generates a voltage pulse in response to an electrical charge generated in the detector 14 by an incident pulse of radiation from a pulsed radiation source 24. The filter circuit allows only frequencies within a predetermined range to contribute to the voltage pulse. The selected range of frequencies is determined in accordance with the duration of the incident radiation pulse such that the voltage pulse represents the magnitude of incident radiation in spite of undesirable detector characteristics which would otherwise introduce distortions. Exemplary charge amplifier and detector circuits are also described.
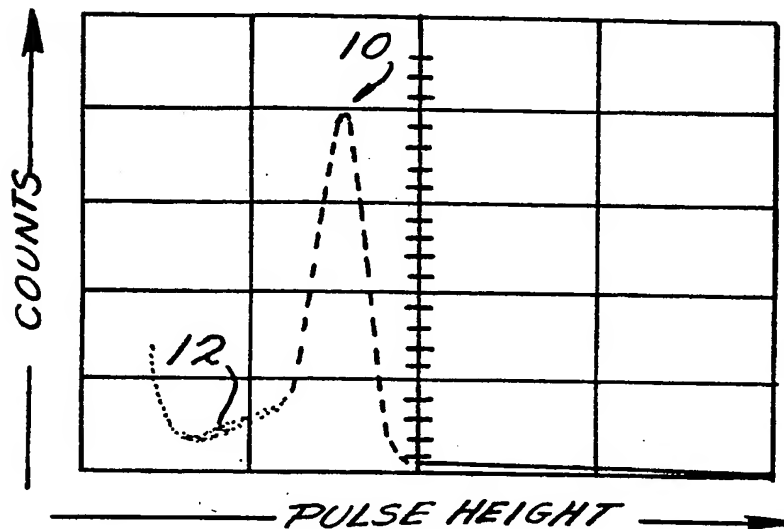

Fig. 4.

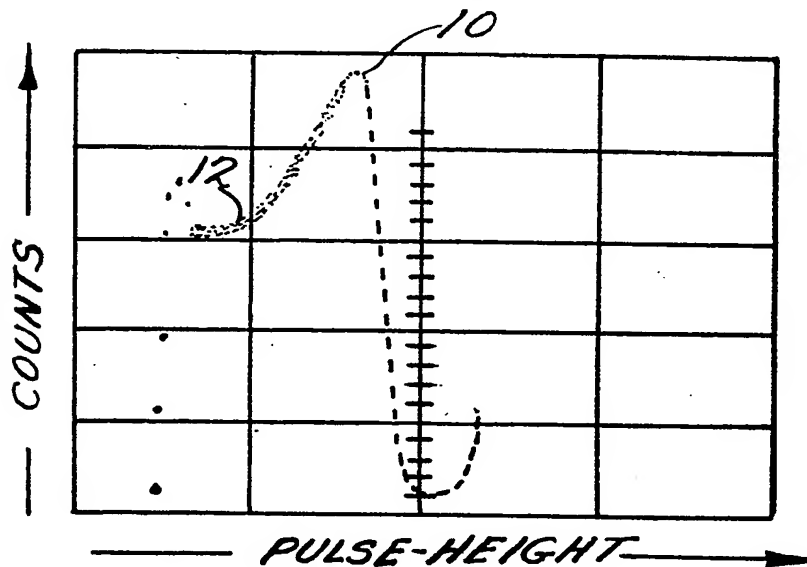# Fig.1A.

COUNTS

PULSE HEIGHT

Fig.1B.

COUNTS

PULSE-HEIGHT

Fig.2.
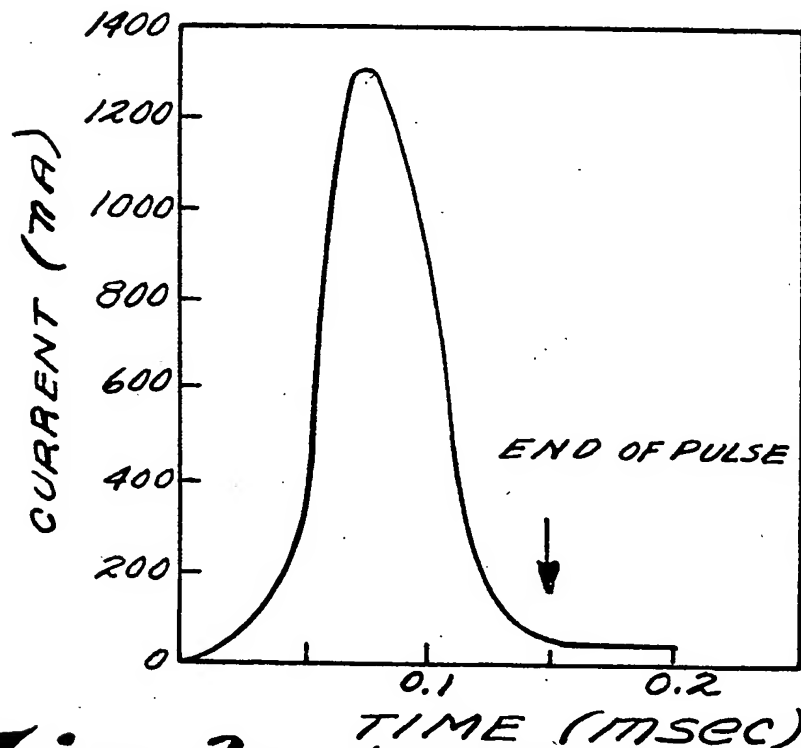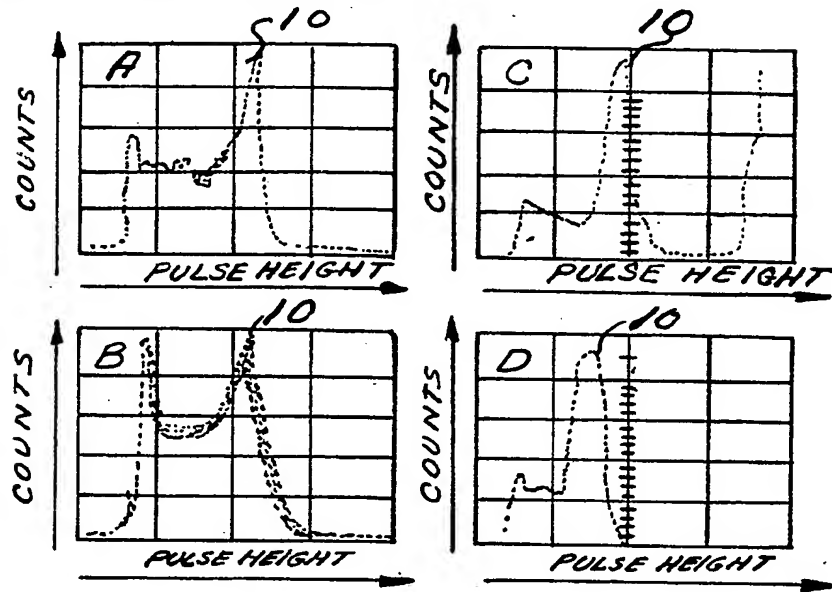
Fig.3a. (PRIOR ART)

Fig.3B.
(PRIOR ART)

CURRENT(mA) vs TIME (m sec.)

END OF PULSE

EXPANDED SCALE

EXPANDED SCALE (mA)

Fig.4.

BIASING CIRCUIT — 16

DETECTOR 14

AC COUPLING 18

Ampl/ FILTER 20

STORAGE AND DATA PROCESSING 22

PULSED RADIATION SOURCE 24

Fig.5.

X-RAY INTENSITY

Tx

TIME (0.1m sec./div)

Fig.6.

Fig. 7.

COUNTS

PULSE-HEIGHT

STOPPING POWER(%) FOR 2mm

$Bi_4Ge_3O_{12}$

CsL

$CdTe$

NaI

CaF_2

xe
$50atm$

Fig.15.

PHOTON ENERGY(KeV)

40 50 60 70 80 100 120 140 150 180

Fig. 8.

(PRIOR ART)

1 cm
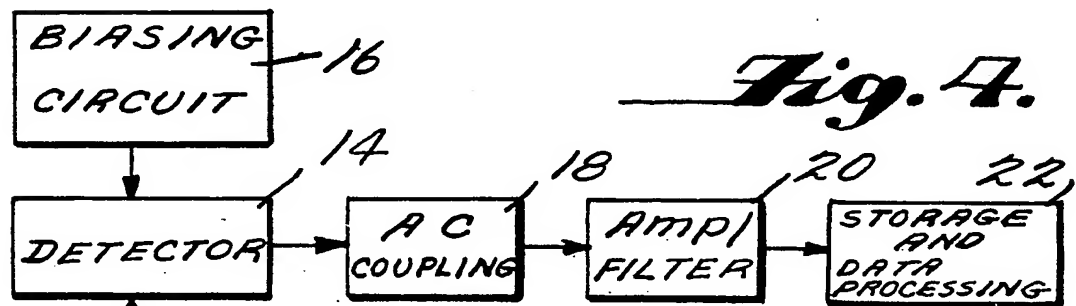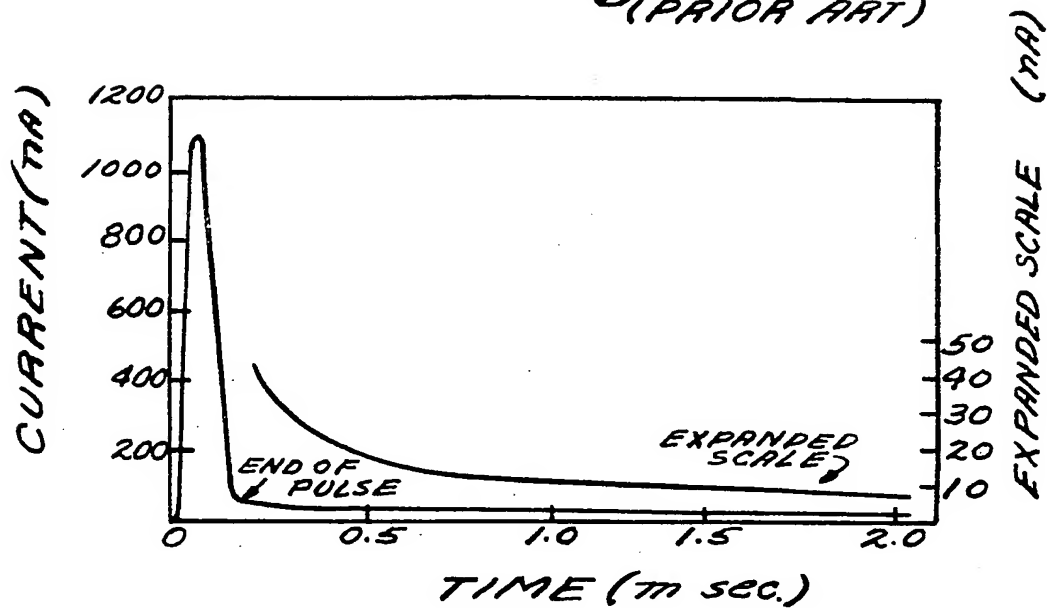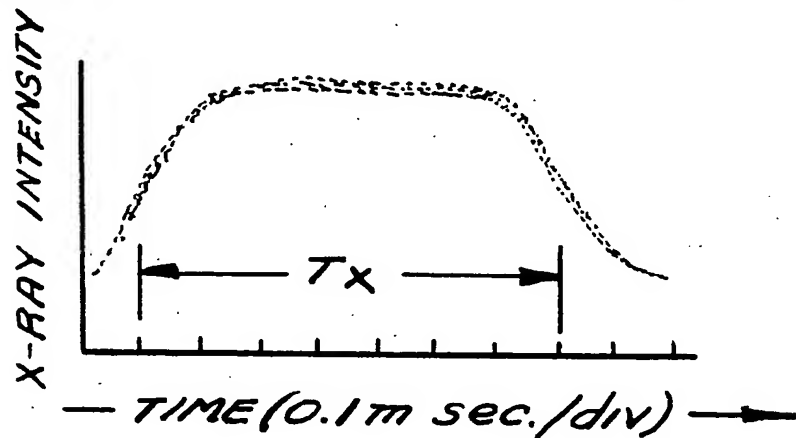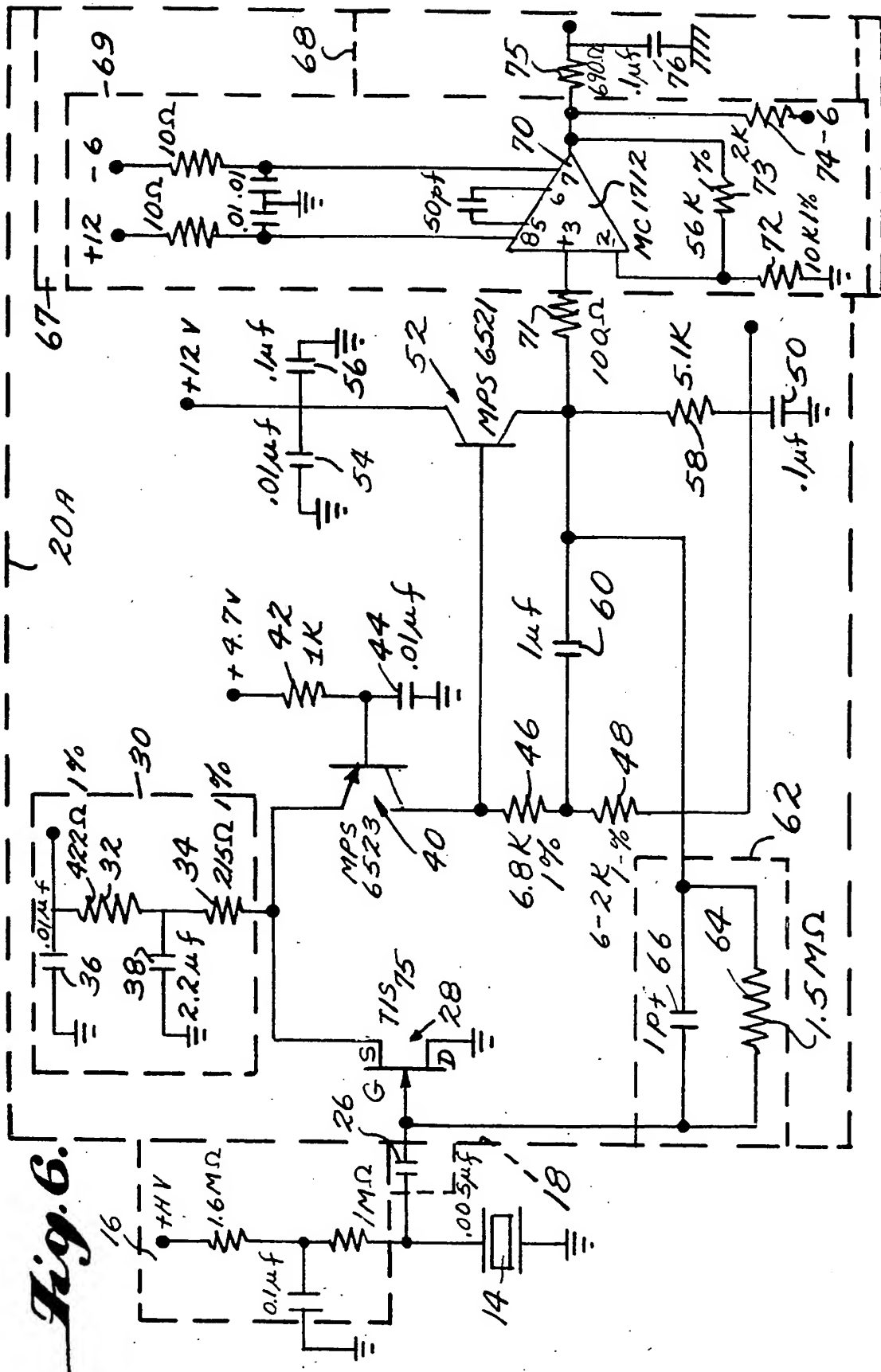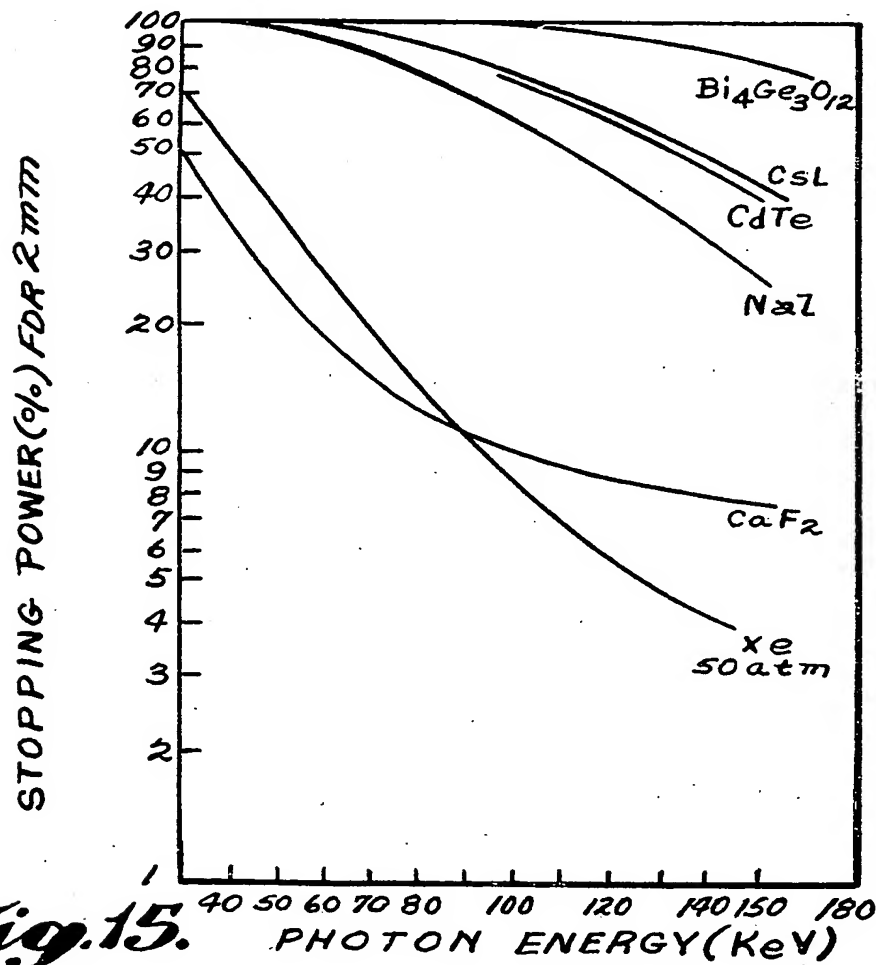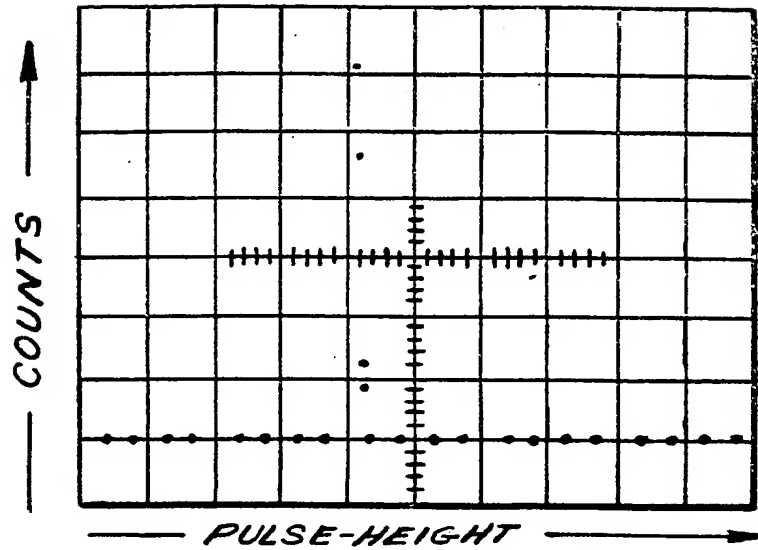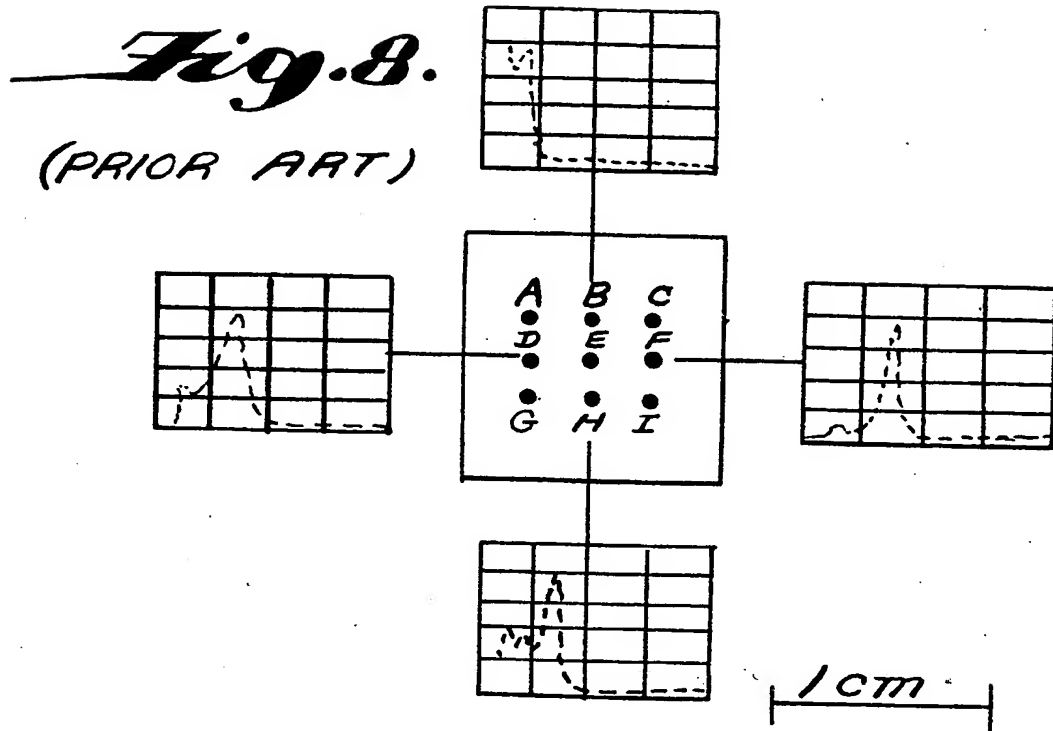
Fig. 9.

PULSED-MODE

POSITION	ENERGY (NORMALIZED)
A	989
B	984
C	825
D	966
E	959
F	1000
G	996
H	982
I	937

9A

9B

Fig. 10A.

Fig. 10B.

Fig.11A.

112
110
14

Fig.11d.

Fig.11b.

114
112d
112f
112c
112b
112e
112a

26

Hv

R=1MΩ
20

Fig.11c.

Fig.12.

CENTER ELEMENT OUTPUT(%)

100

50

10

1

-3 -2 -1 0 1 2 3

SOURCE POSITION(mm)

Fig.13.

A2 COLLIMATED

A6 (STRIP)

AA5 COLLIMATED

AA5 UNCOLLIMATED

DETECTOR OUTPUT (ARBITRARY UNITS)

DETECTOR BIAS (V)

Fig.14.

UNCOLLIMATED

DETECTOR OUTPUT (ARBITRARY UNITS)

DETECTOR BIAS (V)
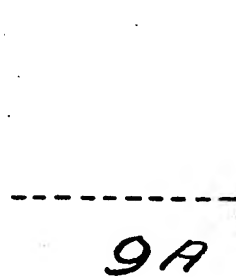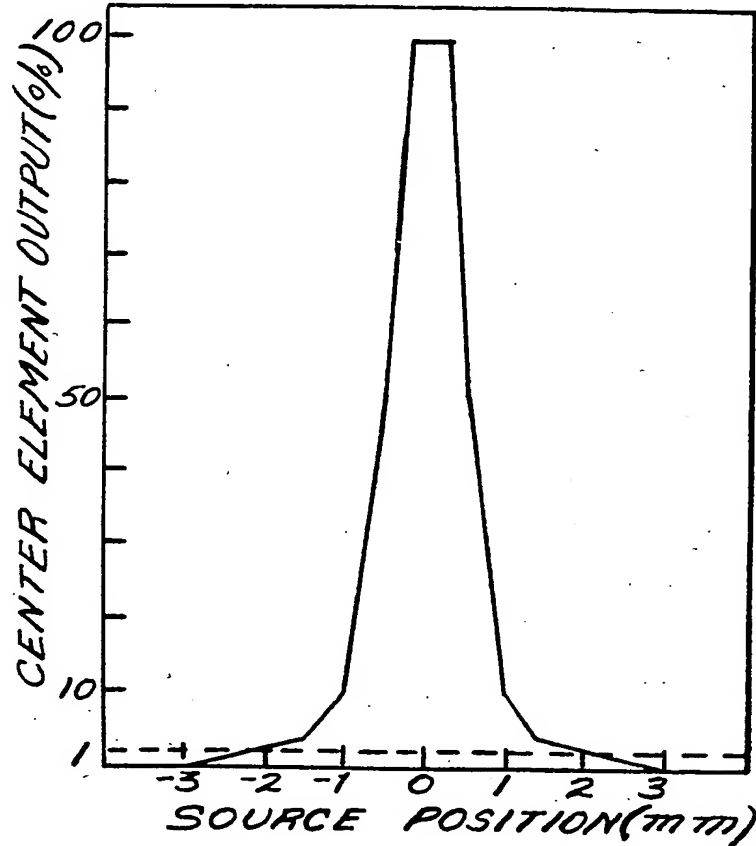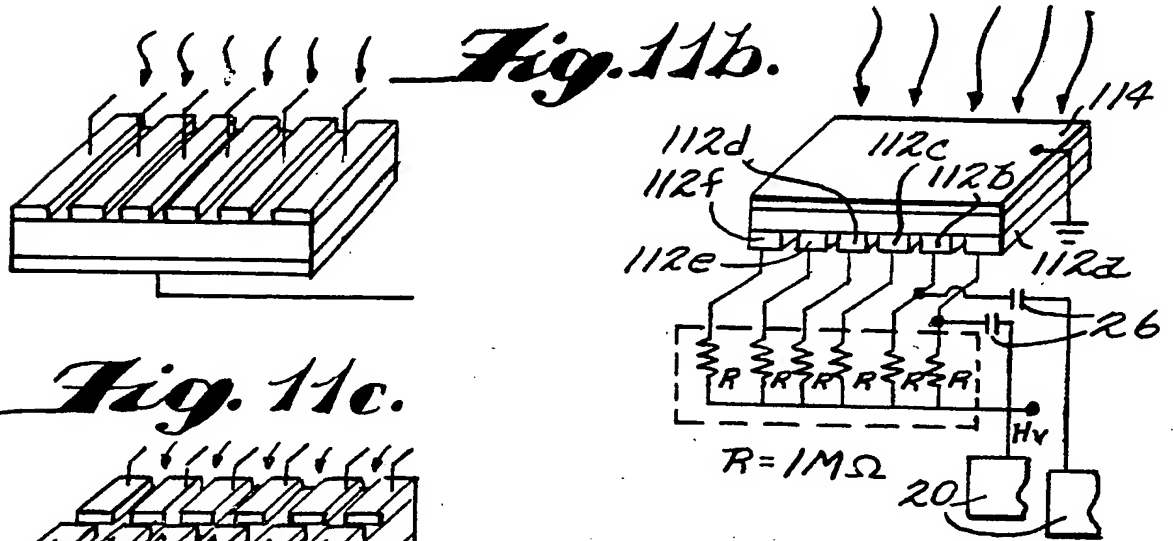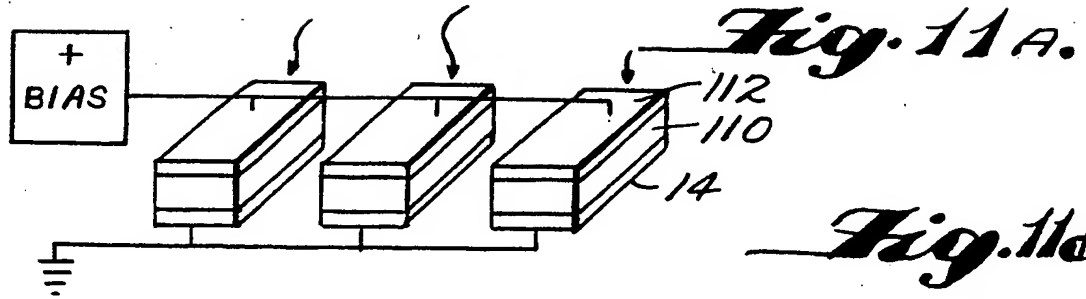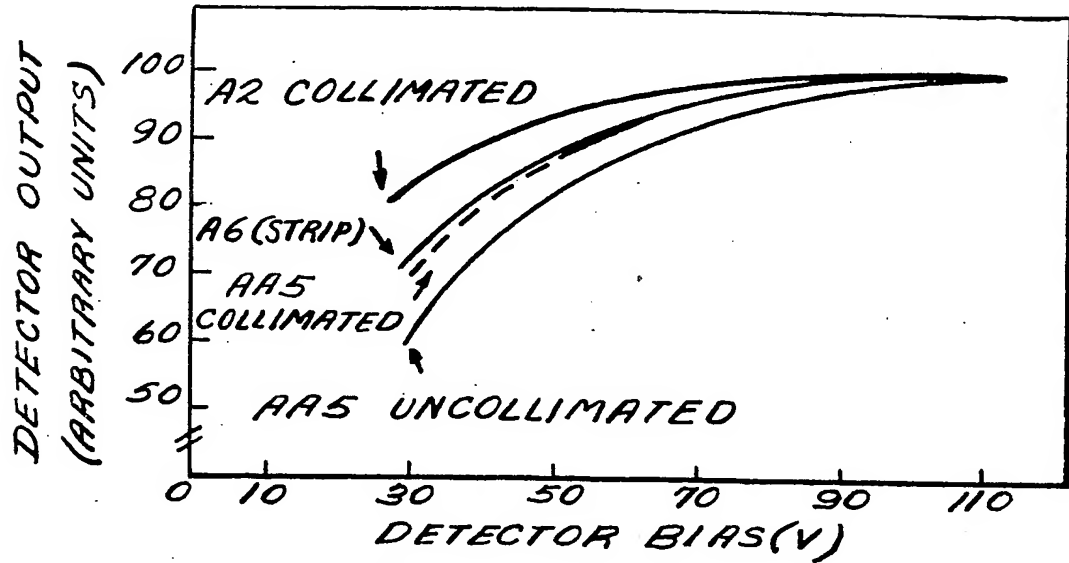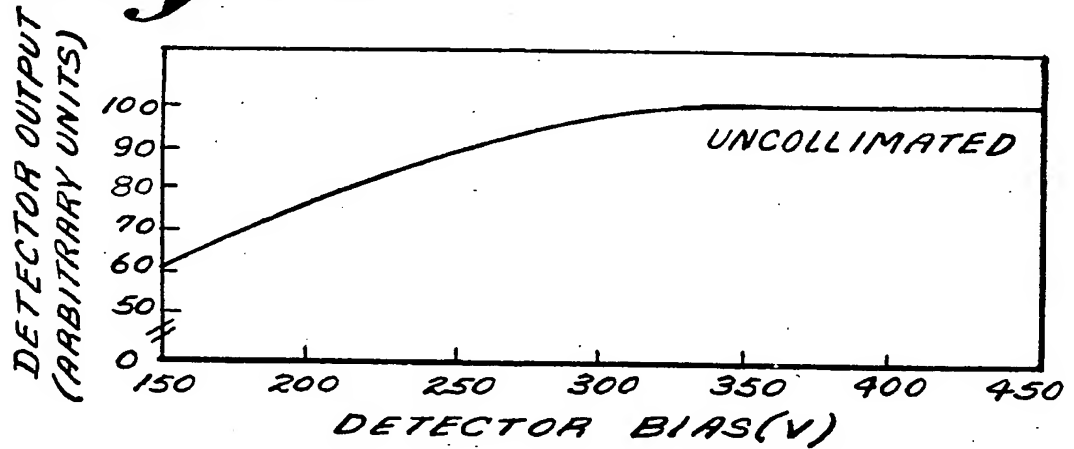
Fig.17.
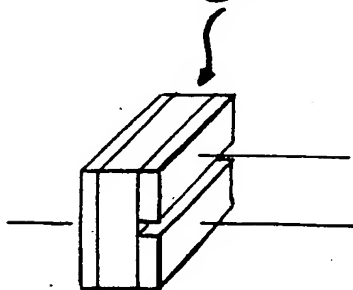
Fig.16.

Fig.18.

SPECIFICATION
Radiation detecting apparatus and method

The present invention relates to a detecting apparatus and to a method of detecting incident radiation pulses, in particular for providing a measurement of radiation transmission/absorption by a
5   cross-section of a body. The apparatus is particularly intended for use in computerised tomographic (CT)   5
X-ray scanning equipment.

There are many devices which now use radiation detectors. For example, CT scanning has proven invaluable for medical diagnosis and analysis and is now in wide use. However, the newest CT scanning techniques tend to be somewhat expensive, due in part to the cost of the components normally
10   associated with a large number of radiation detectors.   10

CT scanners are utilised to provide a computed cross-sectional details of soft living tissue structures. Briefly, the cross-section of interest is positioned between a radiation source (e.g., an X-ray tube) and a detector system. A portion of the beam is absorbed by the tissue during transit along ray paths through the section. Thus, absorption by the body section along any given path is a function of the
15   sum of the absorption coefficients of the particular body tissues through which the beam passes. That   15
portion of the radiation which passes through the section is detected by, for example, a scintillation crystal which produces light photons in response to incident radiation. Typically, the scintillator is optically coupled to a photomultiplier tube (PMT) which converts the light photons into electrical output signals. Such measurements are made along many paths through the body section to provide data used
20   in calculating an array of point-by-point relative absorption coefficients. The computed coefficients are   20
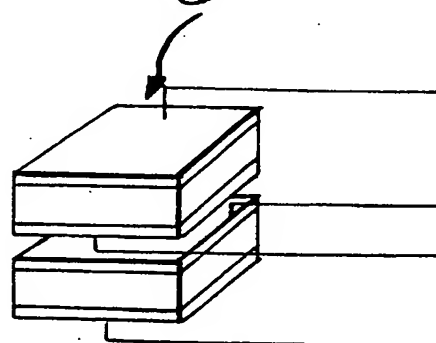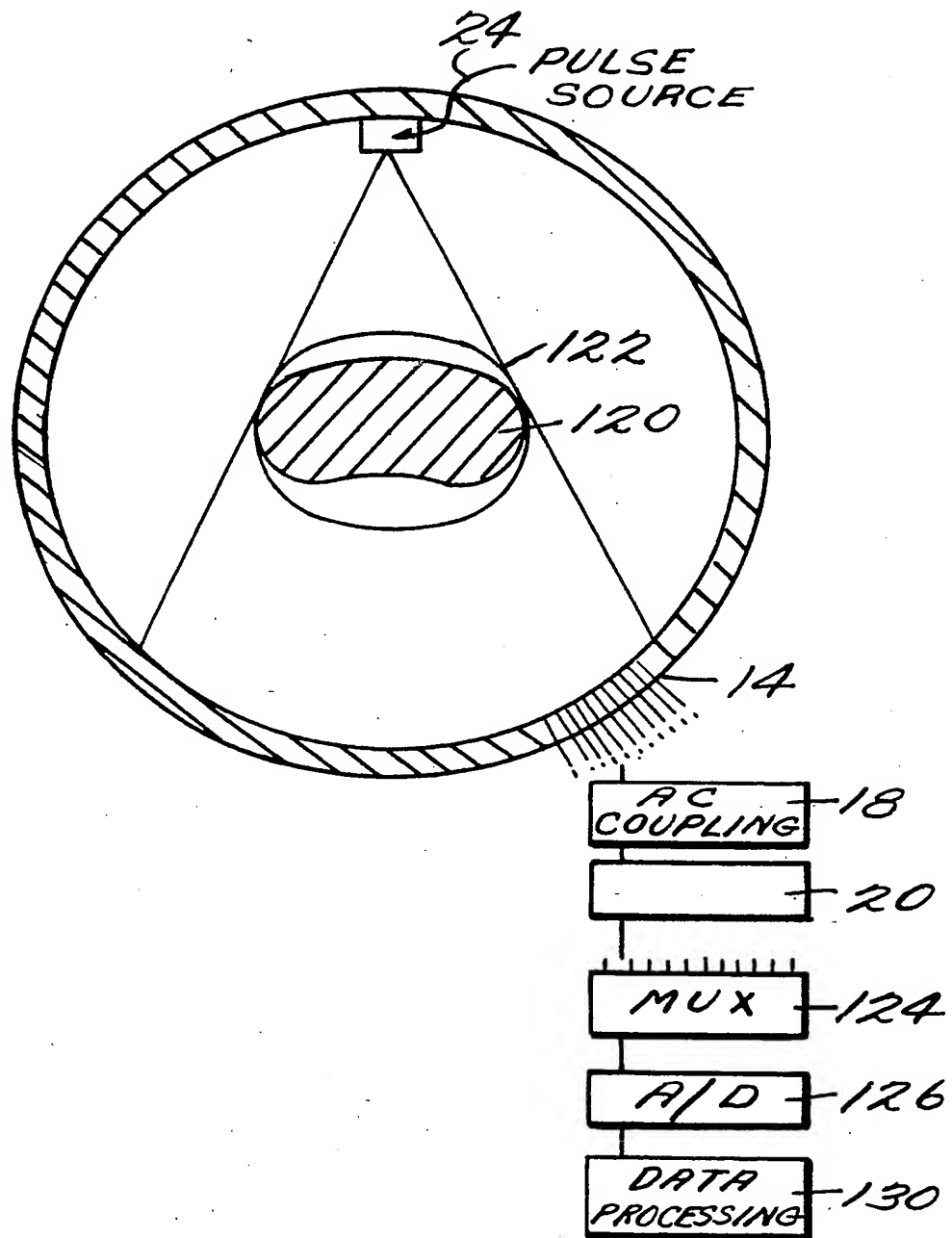then utilised to provide a visual display of the cross-section.

Some of the earlier CT scanners obtain the requisite multiplicity of absorption measurements by synchronously scanning the body section with a highly collimated X-ray beam and an aligned single detector. The beam and detector are together translated to scan the body section and generate a set of
25   measurements along parallel paths. This assembly is then rotated with respect to the body section and   25
the translation-rotation operation is repeated to provide sets of measurements at different angular dispositions.

It is desirable, however, to minimize the overall scanning time. Accordingly, more recent CT scanners generally utilise a plurality of individual radiation detectors in co-operation with a fan-shaped
30   X-ray beam wide enough to irradiate the entire body section. In one example of such systems, a fan-   30
beam source rotates about the body section, irradiates a stationary detector array forming the outside perimeter of the scanning frame. As many as 600 scintillation crystals and associated photomultiplier tubes (PMT) are typically used in such systems. Thus, the cost of the scintillation crystal-PMT detectors, and particularly the photomultiplier tubes, presently represent a substantial portion of the cost of such
35   CT scanners. Accordingly, it is desirable to directly convert the radiation to an electrical signal and   35
eliminate the need for large numbers of photomultiplier tubes and associated circuits.

Various direct conversion detectors such as Xenon gas and high purity germanium (HPGe) semi-conductors have been utilised in the past. However, such detectors are disadvantageous for other reasons.

40   The properties of high purity germanium are well known. A high density position sensitive array   40
has been described in "Two Detector 512-Element High Purity Germanium Camera Prototype" Kaufman et al, IEEE Transactions Nuclear Science, NS-25, February 1978. (Presently in press). However, the production of high purity germanium is a relatively complex and costly procedure, and further high purity germanium detectors require special cooling apparatus, (e.g., liquid nitrogen).
45   Additionally, it is desirable in CT scanning applications to use a thinner detector than generally can be   45
made with high purity germanium.

Room temperature semiconductor detectors would thus appear to be a natural choice for application in CT scanners. Since photo energy from the radiation is directly converted into an electrical charge, compact arrays could be assembled without the requirement of more bulky photomultiplier
50   tubes. Such higher detection element density is especially advantageous for CT scanning applications.   50
These high density semiconductor detector arrays could also be mass produced by batch processing.

However, semiconductor detector materials now available in reasonable quantities have been considered by others in the past and discounted as unsuitable for CT scanning applications. For example, cadmium telluride (CdTe) has specifically been studied with respect to use as a detector in
55   nuclear medicine, and has been found not compatible with the needed accuracies. Specific reference is   55
made to Allemand et al, "Present Limitations of CdTe Detectors in Nuclear Medicine", Revue de Physique Appliquee, 12: 365—367, February 1977. Other semiconductor detectors are discussed in Armantrout et al, "What Can Be Expected from High-Z Semiconductor Detectors", IEEE Transactions on Nuclear Science, Vol. NS-24, No. 1, February 1977. Of the semiconductor materials that are described,
60   various ones have been tested and show unsuitable characteristics similar to CdTe. Computer studies   60
project that similar properties are present in other semiconductor detectors not presently tested.

More specifically, CdTe detectors have been found to exhibit adverse "leakage current", "tailing", "polarisation" and "memory" properties, which are generally considered to render CdTe unsuitable for CT scanning usages. Other detectors, such as mercuric iodide (HgI_2), have been found to exhibit similar

properties.

The high "leakage current" property varies with temperature, and further, varies with time once bias has been applied to the detector. The "tailing" phenomenon causes the detector response to portray a monochromatic (single energy) source as having energies ranging all the way down to noise
5 level. "Tailing" increases as a function of the elapsed time after bias is applied to the detector. The 5 "polarisation" property is manifested by the most frequently encountered detector response (hereinafter termed the photo-peak) to a constant monochromatic source, shifting towards the low energy end of the spectrum. The "memory" property is manifested by a continued current flow after a pulse of incident radiation has ceased, sometimes with a time constant of hours.

10 Various techniques have been processed for dealing with such adverse properties. However, the 10 proposed techniques have tended to merely add to the complexity of the detector circuitry and, in fact, to introduce sources of errors that have made application of CdTe detectors in CT scanners impractical. For example, the high current leakage could be compensated by sampling the detector output just before the X-ray beam is activated to produce each pulse of incident radiation. However, such a scheme
15 will compromise the signal-to-noise ratio of the system when absorption is high, as happens when the 15 beam goes through bone. Pre and post measurement calibration techniques for memory and polarisation compensation, similarly tend to add signal-to-noise problems. Further, polarisation compensation techniques wherein the bias to the detectors is turned off between measurements, are inadequate due to hysteresis effects.

20 In accordance with a first aspect of the present invention, there is provided a radiation detecting 20 apparatus for use in detecting incident radiation pulses of predetermined duration, said apparatus comprising a detector which generates electrical charge in response to incident radiation, said detector being disposed to receive at least a portion of said radiation pulses; signal processing means AC coupled electrically to said detector for generating an output electrical signal accurately representing
25 said incident radiation, said signal processing means including filter means for allowing only frequency 25 components within a predetermined range of frequencies to contribute to said output electrical signal, said range of frequencies being determined in accordance with the duration of said incident radiation pulses so that said output electric signal is undistorted by variations in the internal electrical characteristics of the detector. Preferably the detector is a semiconductor type.

30 It has been found that room temperature semiconductor detectors, and in particular CdTe 30 detectors, can be made compatible with CT scanning operations. The detector is preferably used in co-operation with a pulsed radiation source and the frequency filtering techniques are utilised in measuring the resulting pulse of electrical charge developed by the detector. Only selected frequencies within a predetermined range, chosen in accordance with the duration of incident radiation pulse, contribute to
35 the output pulse. Preferably, such range of frequencies is preferably a narrow band of frequencies 35 centred about a frequency approximately equal to the inverse of twice the radiation pulse duration.

In a preferred embodiment, a plurality of individual detectors are formed as a unitary array on a single CdTe or h_gI_2 wafer by merely providing plural electrically separated electrodes. Thus, CdTe or H_gI_2 detectors are particularly advantageous in that entire high density arrays can easily and quickly be
40 formed by batch processing. Further, two dimensional arrays can be formed by providing a grid work of 40 such electrodes. Moreover, a stacked array for taking measurements at a plurality of differing energy levels can be provided by disposing a multi-electrode detector such that the radiation is incident on a wafer surface transverse to the electrodes.

In accordance with a second aspect of the invention, there is provided a method of detecting
45 incident radiation pulses of predetermined duration, said method comprising the steps of disposing a 45 semiconductor detector to receive at least a portion of said radiation, said semiconductor detector generating an electrical charge in response to said received radiation; generating an electrical output signal indicative of said charge; and filtering out components of said electrical output signal not in accordance with the duration of said radiation pulses so that said electrical output signal is substantially
50 undistorted by the characteristics of said semiconductor detector. 50

In accordance with a third aspect of the invention, there is provided a method for providing a measurement of radiation transmission-absorption by a cross-section of a body, comprising the steps of, generating pulses of radiation of predetermined duration, directing said radiation pulses to irradiate said body, disposing a semiconductor detector to receive at least a portion of said radiation transmitted
55 through said body, said semiconductor detector generating an electrical charge in response to said 55 received radiation, generating an electrical output signal indicative of said charge, and filtering out components of said electrical output signal not in accordance with the duration of said radiation pulses so that said electrical output signal is substantially undistorted by the characteristics of said semiconductor detector.

60 In order that the invention may be better understood, several embodiments thereof will now be 60 described by way of example only and with reference to the accompanying drawings, in which:—

Figures 1A and 1B are spectral responses of CdTe detectors when used as in the prior art illustrating the phenomenon of "tailing";

Figures 2A—2D are spectral response of CdTe detectors when used as in the prior art illustrating
65 the phenomenon of "polarisation"; 65

Figures 3A—3B are graphs illustrating "memory" of CdTe detectors when used as in the prior art;

Figure 4 is a block diagram of a radiation detecting apparatus in accordance with the present invention;

Figure 5 represents the shape of a typical X-ray pulse;

5 Figure 6 is a schematic diagram of a suitable charge amplifier and filter means for use in the apparatus of Figure 4;

Figure 7 is the spectral response of a CdTe detector used in accordance with the present invention;

Figure 8 illustrates the non-uniform response of a CdTe detector as used in the prior art;

10 Figures 9A and 9B illustrate the uniform response of the CdTe detector of Figure 8 when used in accordance with the present invention;

Figures 10A and 10B are schematic diagrams of alternative charge amplifier and filter means;

Figures 11A—11D are pictorial schematics of CdTe detectors in accordance with another embodiment of the present invention;

15 Figure 12 illustrates the localized charge phenomenon of CdTe, and lack of crosstalk between elements in accordance with this invention;

Figures 13 and 14 illustrate the charge response of CdTe detectors to bias voltage;

Figure 15 compares the radiation stopping power of various detector materials;

Figures 16 and 17 illustrate dual-energy detectors; and

20 Figure 18 is block diagram of a CT scanner utilising detectors in accordance with the present invention.

The properties of CdTe detectors have been examined by observing the spectral response to single photons from a monochromatic source (AM-241). Figures 1A and 1B illustrate the spectral response of two different CdTe detectors to a monochromatic source (AM-241) when used as in the prior art.

25 Specifically, Figures 1A and 1B show the count of photons registered by the detector at various energy levels during a preset time interval, the energy level being indicated by the pulse height of the detector output.

Ideally, since the response was to a monochromatic source, all of the counts should have registered at a single energy providing a single impulse-like spike response. However, as illustrated in

30 Figure 1A, due to the tailing effect, the spectral response is broadened. Photons which should be depicted as counts at the photo-peak 10 (predominant energy level) are indicated as having energies ranging down to the noise level of the detector, and a plateau region 12 is manifested in the spectral response.

It is apparent from a comparison of Figures 1A and 1B, that the tailing effect is not uniform

35 between individual detectors. The response shown in Figure 1A is that of a relatively good (by prior art standards) CdTe detector, whereas the response shown in Figure 1B of another CdTe detector to the same source evidences a worse peak-to-plateau ratio and poor energy resolution.

The polarisation phenomenon shifts the photo-peak response towards the lower energy levels and the peak-to-plateau ratio decreases (a measure of the tailing effect) as a function of the time from the

40 instant the detector is placed under bias as illustrated in Figures 2A—2D. Figures 2A and 2C show the spectral response of two different CdTe detectors to monochromatic AM—241 immediately after bias is applied. Figures 2B and 2D illustrate the spectral response of the respective detectors 15 minutes after bias is applied. It should be noted that the photo-peak 10 shifts towards the lower energy level, and the response widens as tailing increases. Such polarisation has generally been attributed (apparently in

45 error, as will be explained) to the loss of depletion depth with time. If such is the case, it follows that X-ray detection efficiency would be reduced as a function of time and since tailing worsens with time, the charge collection efficiency of the detector would also degrade with time. In any event, irrespective of the physics involved in an uncompensated detector, a steadily decreasing response to a steady X-ray source would be manifested.

50 Figures 3A and 3B illustrate the memory property exhibited by CdTe detectors with current continuing to flow after an applied radiatioin pulse has terminated. More specifically, the current output of two different CdTe detectors are shown respectively in Figures 3A and 3B as a function of time, following exposure to a short X-ray pulse. It should be noted that leakage current does not decrease below 1% of maximum until approximately 25 μsec. and 800 μsec. after the end of the X-ray pulse for

55 the respective detectors associated with Figures 3A and 3B. DC relaxation times were found to be approximately 0.5 minutes and 1.5 minutes respectively. Such memory phenomenon, absent compensation, would be unusable with CT scanners at present scanning rates.

Data is available which tends to indicate that the tailing effect is the result of the CdTe detector producing pulses of varying rise times. Reference in this regard is made to Jones, "the Use of CdTe

60 Spectrometers in Monitoring Activity Deposited in Nuclear Power Stations", Revue de Physique Appliquee 12: 379—384, 1977. When processed with conventional amplifiers having Gaussian-shaped frequency responses, the varying rise times are manifested as pulses of different amplitudes. The above-noted Jones article appears to demonstrate that the fastest pulses correspond to photons of the highest energy level registered (the photo-peak) and that various bands in rise time correspond to

65 energy bands in the tail.

While the polarisation phenomenon is generally considered to be the result of a decreasing depletion depth, as noted above, the present inventor speculates that the polarisation effect is the result of further degration of rise time distributions during the period after bias is applied. Accordingly, polarisation and tailing effects could effectively be eliminated if the detector is presented with a pulse of
5 photons having a duration which is long compared to the longest detector rise times (approximately 5 1 μsec.). Thus, the pulse of radiation in effect drives the rise time response, and spurious signals not in accordance with the rise time of the pulse can be filtered from the detector output to provide an undistorted representation of the incident radiation pulse. Experimental results appear to verify this theory. In any event, irrespective of the physics of the phenomenon, the experimental results have
10 proved that semiconductor detectors can be made compatible with the requisites of CT scanning when 10 practiced in accordance with the techniques described herein.

With reference now to Figure 4A, a semiconductor detector 14 such as CdTe, suitably biassed by circuitry 16, is AC coupled via means 18, (e.g., a capacitor), to an amplifier/filter 20. Amplifier/filter 20 is coupled to suitable storage and/or data processing means 22.
15 Detector 14 cooperates with a pulsed radiation source 24. Source 24 can be a conventional pulse 15 mode X-ray generator, or can be a constant source such as, for example, a Phillips MG160 X-ray generator, used in conjunction with a wheel chopper. Where wheel choppers are used, the rotational frequency of the wheel and ripple in the X-ray tube output signal tend to produce beats in pulse amplitude. Such beat frequencies can be compensated for by averaging a number of pulses. A typical X-
20 ray pulse shape is shown in Figure 5. 20

The effective time constant of biassing circuitry 16 and AC coupling 18 is chosen in accordance with the pulse repetition rate of source 24; the time constant being chosen to allow detector 14 to recharge (recover) between radiation pulses.

Amplifier/filter 20 comprises means for generating a signal indicative of the charge developed by
25 detector 14, and has a frequency response such that frequency components not within a predetermined 25 range (band) of frequencies are effectively filtered from that signal. The predetermined frequency band is determined in accordance with the radiation pulse duration and is preferbly a narrow band of frequencies centred about a frequency approximately equal to the inverse of the radiation pulse duration or the inverse of twice the radiation pulse duration. For example, amplifier/filter 20 suitably comprises a
30 charge sensitive first stage having a RC time constant TD, and a second stage having an effective RC 30 time constant TS. It was found that by choosing the time constants such that TS is approximately equal to the duration of the X-ray pulse (TX) and much larger than TD, a strongly differentiated detector output pulse results, which provides faithful representation of the actual X-ray pulse shape. Accordingly, the integration of such a differentiated "X-ray pulse" yields an accurate indication of the total incident X-ray
35 energy. Such an amplifier/filter will hereinafter be described in more detail in conjugation with Figure 6. 35

In the alternative, the charge sensitive first stage can be made to have an RC time constant much larger than the duration TX of the X-ray pulse, and the second stage effectively shaping the first stage output by differentiation with time constant TD and integration with a time constant TS considerably shorter than TD but comparable to TX. Such a filtering amplifier (20B) will be more fully described in
40 conjunction with Figures 10A and 10B. While such an arrangement successfully eliminates the effects 40 of the detector characteristics from the output pulse it is often difficult to obtain adequate pole-zero compensation and tends to exhibit undesirable overshoot characteristics.

With reference now to Figure 6, amplifier/filter 20A will now be described. Exemplary component values are given for use in conjunction with X-ray pulse durations on the order of 100μsec.
45 A CdTe detector 14 is biassed by a biassing circuit 16, and coupled through AC coupling means 45 18 to amplifier/filter 20A. AC coupling means 18 may be a 0.005 μf. capacitor 26. Capacitor 26 is coupled to the gate terminal of a field effect transistor 28, suitably of the type TIS 75, having grounded source and a drain suitably biassed by a resistor-capacitor biassing network 30. Biassing network 30 may comprise resistor 32 (422 Ω) and 34 (215 Ω) serially connected between the source of FET 28 and
50 a 24 volt source. Bypass capacitors 36 (0.1 μf.) and 38 (212 μf.), are respectively coupled from the 50 voltage source to ground and from the juncture between resistors 32 and 34 to ground.

The drain of FET 28 is also connected to the emitter of an PNP transistor 40 (e.g., of the type MPS 6523). The base of transistor 40 is coupled through a resistor 42 to a positive 4.7 volt source, and is AC coupled to ground through a capacitor 44 (0.01 μf.). The collector of transistor 40 is coupled through
55 two serially connected resistors 46 and 48 to a negative 12 volt source. An AC bypass capacitor 50 is 55 provided for the negative 12 volt source.

The collector of transistor 40 is also coupled to the base of an NPN transistor 52, (e.g., of the type MPS 6521). The collector of transistor 52 is connected to a positive 12 volt supply, the 12 volt supply being coupled to ground through each of two capacitors 54 (0.01 μf.) and 56 (0.1 μf.), respectively. The
60 emitter of transistor 52 is coupled to the negative 12 volt source through a resistor 58 (5.1KΩ). The 60 emitter of transistor 52 is also connected through a capacitor 60 (1μf.) to the juncture between resistors 46 and 48, (connected between transistor collector 40 and the negative 12 volt supply).

Additionally, a fast differentiator 62 is connected in a feedback loop between the emitter of transistor 52 and the gate of FET 28. Differentiator 62 may comprise a resistor 64, (1.5MΩ), in parallel
65 with a capacitor 66, (1pf.). 65

The emitter of transistor 52 is also coupled to a second stage 67, comprising an integrator 68 preceded by a voltage amplifier 69. Voltage amplifier 69 suitably comprises a conventional operational amplifier (op amp) 70, for example, of the type MC1712. The emitter of transistor 52 coupled to the non-inverting input terminal of op amp 70 through a precision resistor 71 (100Ω). The inverting input
5 terminal of op amp 70 is connected to ground potential through a precision resistor 72 (10KΩ), and feedback between the output terminal and inverting input terminal of op amp 70 is provided by a further precision resistor 73 (56KΩ). A connection is also made from the output terminal of op amp 70 to a negative 6 volt source through a resistor 74 (2K) to provide for generation of bipolar signals by amplifier 69. The output terminal of voltage amplifier 67 is connected to integrator 68, suitably comprising a
10 resistor 75 (690Ω) and a capacitor 76 (0.1 μf).

The circuit of Figure 6 can be adapted for radiation pulse durations other than on the order of 100μs, by adjusting the time constants of differentiator 62 and integrator 68. For example, the circuit can be made compatible with a 800μs radition pulse duration by adjusting the values of resistor 75 and capacitor 76 to increase the time constant of integrator 68 by a factor of 8 to 10.
15 A number of different detectors were tested in conjunction with the circuit of Figure 6. The structure of such detectors will now be discussed below. For comparison, the uncompensated signal photon characteristics of the detectors (as used in the prior art) were first noted. The results of the testing are shown in tabular form in the accompanying Tables I—V. Table I is a brief description of the individual detectors tested. The detector structures, *per se,* are conventional devices, except that in
20 some instances the detectors are adapted, as will be explained, for operation with plural electrodes to form an array. Tables II and III show the uncompensated characteristics of the detectors. Table II shows the detector characteristics when bias voltage is first applied at the detector, and Table III shows the measurements taken 15 minutes after bias is applied.

TABLE 1

DETECTOR SPECIFICATIONS

Detector Code	Detector Model No.	Description	Electrodes
A1	Radiation Monitoring Device (R.M.D.) NRD 0201	Rectangular 2mm × 7mm × 2mm	Platinum (Pt)/Pt
A2	M272—2	Round (single electrode) 8mm Dia. × 2mm	Pt/Pt
A3	M141—4	Round (single electrode) 10mm Dia. × 2mm	Pt/aluminium-oxide 0—A1 (Pt. side neg. (Purposely fabricated for poor results)
AA4	ND040—01	Round (split electrode) 10mm Dia. × 2mm	Pt/Pt
AA5	ND040—02	Round (split electrode) 10mm Dia. × 2mm	Pt/Pt
A6	M353—4	Rectangular 10mm × 15mm × 2mm (multi electrode)	Pt/Pt
B7	Hughes CTD—UCSF—1	10 × 10 × 2mm³	Aquadhag/ Aquadhag

TABLE II

DETECTOR CHARACTERISTICS

(Single Photon Mode)

Detector Code	Bias Voltage	Leakage Current	Energy Resolution Full-width-at-half-maximum-height FWHM (at 60 Kev)	Peak to Plateau
A1	60V	35 nA	23.1%	2.68
A2	60V	500 nA	42.3%	2.75
A3	50V	14 nA	19.3%	7.28
AA4 Side 1	80V	300 nA	—	2.19
Side 2	60V	150 nA	—	2.05
AA5 Side 1	60V	90 nA	—	1.69
Side 2	60V	77 nA	—	1.66
A6	50V	120 nA per element	—	—
B7	350V	38 nA	24.8%	5.38

TABLE III

DETECTOR STABILITY

(Single Photon Mode)

Detector Code	Leakage* Current Change	Photopeak* Shift (at 60 keV)	Peak to Plateau* Change (at 60 keV)
A1	35—50 nA	−0.56 keV	2.68 → 2.67
A2	no change	no change	no change
A3	no change	−10.5 keV	7.28 → 4.22
AA4 1	breakdown		
2	220 → 260 nA	no change	2.05 → 1.98
AA5 1	90 → 110 nA	−1.6 keV	1.69 → 1.61
2	77 → 89 nA	−0.8 keV	1.66 → 1.58
A6	120 → 170 nA	not measured	not measured
B7	38 → 40 nA	no change	5.38 → 4.92

* Change over 15 minutes after the detector is under bias.

As shown by Tables I—III the detectors were found to have initial leakage currents ranging from 14nA to 500nA, energy resolutions from 19.3% to 42.3% and peak-to-plateau ratios (see Figure 1) ranging from 7.28 to 1.66. Subsequent measurements showed that 15 minutes after the detectors were placed under bias, changes in leakage currents ranging from 0 to 50nA were evidenced, shifts in photo-peak ranged from 0 to −10keV and the peak-to-plateau ratio evidenced changes ranging from 0 to almost 3.

The detectors were then tested in pulsed X-ray mode, in conjunction with amplifier/filter 20A (Figure 6), using X-ray pulse durations of approximately 100μsec. and 800μsec. Each of the detectors

when used in cooperation with the amplifier 20A (Figure 6) provided output signals with spectral
responses such as that shown in Figure 7, approaching the ideal impulse spectral response with no
appreciable memory or tailing effects manifested.

5   The memory effects were tested by measuring the amplitude of a small pulse that was interlaced
with a large pulse (approximately four times the amplitude of the smaller pulse). A wheel chopper was   5
utilized to produce a pulsed beam with pulses occurring at intervals of approximately 20msec.,
equivalent to the rate necessary for obtaining 1° increments of measurement in rotary CT scanner
operating at a speed such that a 360° rotation is completed in 7.2 seconds. This large pulse was
blocked with a synchronous second chopper to provide for observation of the change in measured

10   amplitude of the smaller pulse. If the large memory effect observed in the uncompensated detector   10
existed, a large shift would be manifested in the representation of the amplitude of the smaller pulse.
However, no measurable shift was observed within a 1% measurement limit. Measurements were taken
at 3.5 minute intervals over a period of 35 minutes. The results are summarized in Tables IV and V. The
results demonstrate that after a short turn-on period, stable operation was attained by utilizing the

15   pulsed source and circuit shown in Figure 6, for measurement periods over approximately half an hour.   15

TABLE IV

DETECTOR STABILITY

(Pulsed Mode)

Detector Code	Time to Stable Response	% Shift From Time Zero	Pulsed Width Used	No. of Pulses Per Event	Pulse % FWHM
A2	3.5 min.* 3.5 min.	4.6% 2.9%	~100μs ~800μs	20 20	2.5% 2.8%
A3	10.5 min. 10.5 min.	6.2% 6.7%	~100μs ~800μs	20 20	4.0% 3.8%
AA5	7.0 min.	1.9%	~100μs	20	2.8%
A6	3.5 min.*	0.5%	~800μs	10	4.6%
B7	3.5 min.*	0.7%	~100μs	20	3.6%

*NOTE: Measurements made at 3.5 min. intervals.

TABLE V

DETECTOR STABILITY

(Pulsed Mode)

TIME			OUTPUT	(± .010)			
(min.)	A2 *	+	A3 *	+	AA5 *	A6 +	B7 *
0	1	1	1	1	1	1	1
3.5	.954	.971	.967	.973	.997	.993	.995
7	.960	.970	.945	.933	.982	.993	.999
10.5	.969	.974	.938	.917	.988	.992	.978
14	.970	.972	.937	.896	.987	.991	.979
17.5	.975	.968	.937	.876	.985	.995	.990
21	.984		.938	.855	.977	.981	.997
24.5	.982			.849	.980	.971	.997
28	.985			.857	.986	.979	.992
31.5	.982				.981	.980	.973
35	.984					.990	.984

*100 sec. pulse width

One of the problems with prior art usage of semiconductor detectors in general, and CdTe in particular, is the non-uniformity of the response exhibited between respective detectors, and, in addition, non-uniformity of response over the surface of a respective detector. The uniformity of response of the detectors when used in conjunction with a pulsed source and amplifier/filter 20 was
5 checked, by collimating the X-ray source to a small area of the detector.

Referring now to Figures 8 and 9, localised measurements were taken with the collimated radiation directed to various positions A-1 on the detector surface. Such positions are illustrated in Figure 8, along with the spectral responses to a monochromatic source (AM-241) at various of those positions for the case of the most non-uniform detector (B7 in Tables I—V) when used in accordance
10 with the prior art. Figure 9B shows the spectral response found at each of the positions A-1 when the detector B7 was used in conjunction with a pulsed X-ray source and filtering amplifier 20. Figure 9A more particularly indicates the normalised photo-peak response of the CdTe detector to collimated radiation at positions A-1 using this invention.

It should be noted that while the CdTe detector has extremely non-uniform response when used in
15 accordance with the prior art techniques, the deviations in response are of the order of $\pm$ 2% when used in accordance with the present invention. More significantly, it was found that even using individual detectors of different geometry, (and in fact, made by different manufacturers) when the radiation source was collimated to provide equal irradiated areas, these detectors provided readings within 7% of each other when used in accordance with the present invention.
20 As noted above the amplifer/filter 20B depicted in Figure 10A also successfully compensates for the characteristics of detector 14. The exemplary component values here given are again suitable for operating with X-ray pulse durations on the order of 100 sec. In the circuit of Figure 10, FET 28 and transistors 40 and 52 are interconnected in the same manner as in Figure 6, with the exception that fast differentiator 62 is replaced by long time constant circuit 80 comprising a parallel combination of a
25 resistor 82 (120MΩ), and capacitor 84 (10pf.).

The emitter of transistor 52 is coupled to an active second state 85. Active stage 85 comprises a differentiator portion 86 comprising an operational amplifier (op amp) 88, (e.g., of the type MC1712) and an integrator portion 100. The emitter of transistor 52 is coupled through a precision resistor 90, (100μ) to a non-inverting input of op amp 88. The inverting input of op amp 88 is coupled through a
30 serially connected precision resistor 92 (100Ω) and capacitor 94 (0.01μf.) to ground. Feedback from the output terminal of amplifier 88 to its inverting input is provided through a further precision resistor 96, (6.19KΩ). A connection from the output terminal of op amp 88 is made through a resistor 98 (2KΩ) to a —6 volt source. The output terminal of op amp 88 is also connected to an integrator 100 comprising a resistor 102 (690Ω) and a capacitor 104 (100pf.). Amplifier/filter 20B can be adapted for other
35 radiation durations by changing the time constant of differentiator 86, integrator 100, and, where necessary, circuit 80. Similarly, for long pulse durations it may be necessary to adjust the effective time constant of biassing circuit 16 and coupling circuit 18.

Figure 10B depicts an amplifier/filter 20C similar to amplifier/filter 20B, but including provisions for pole zero compensation. The exemplary values given are again associated with a radiation pulse
40 duration of the order of 100μs.

The long time constant circuit 80 of amplifier/filter 20B is replaced in amplifier/filter 20C by a long time constant circuit 80a of slightly different time constant, formed of a parallel combination of resistor 82a (100MΩ) and capacitor 84a (18pf). The emitter of transistor 52 is coupled through two serially connected capacitors 105 (1 μf) and 106 (0.1 μf) to the input of a voltage amplifier 107, suitably identical to
45 voltage amplifier 67 in Figure 6. A potentiometer 108 (5K) is connected between the juncture of capacitors 105 and 106 to ground potential. The wiper of potentiometer 108 is connected through a precision resistor 109 (10K) to the input of voltage amplifier 107. Capacitor 106 and resistor 109 cooperate to form an effective differentiator having a time constant of 1ms.

Resistor 109 and capacitor 105 also cooperate with capacitor 105 and potentiometer 108 as a
50 pole zero (p/z) compensation circuit.

The output of voltage amplifier 107 is connected to an integrator 100a, comprising a resistor 102 (690Ω) and capacitor 104a (0.1 μf). The time constant of integrator 100a is thus approximately 70 μs.

It should be appreciated that any of amplifier/filters 20A, 20B or 20C can be made compatible with faster pulse repetition frequencies by selectively discharging the capacitor (76, 104, 104a) of the
55 integrator (68, 100, 100a) after the peak charge developed by detector 14 in response to a given radiation pulse has been attained. This can be easily accomplished by shunting the conduction path of a transistor across the capacitor (76, 104, 104a) to ground, and controlling conduction through the transistor in accordance with the output signals of a conventional peak detector responsive to the output signal of the integrator.
60 As noted above, a conventional semiconductor detector such as CdTe or H$_g$I$_2$ can be made compatible with the requisites of CT scanning in accordance with the above described aspect of the present invention. As illustrated in the Figure 11A, conventional semiconductor detectors comprise a wafer of the semiconductor material, (e.g., CdTe 110) having deposited on either side thereof conductive electrodes 112 and 114. The electrodes are typically formed of platinum or aquadhag.
65 Such unitary detector arrays are shown in Figures 11B and 11C. Assuming the incident radiation

to scan across the detector in the longitudinal direction, an array of adjacent detectors can readily be provided by merely removing strips of electrode material in the transverse direction at longitudinally spaced locations to provide a plurality of electrically isolated electrodes Figure 11B. Similarly, by additionally removing a strip of electrode material in the longitudinal direction (Figure 11C) a two-dimensional array for simultaneously detecting multiple cross-sections in a CT scanner is provided.

In Figure 11D a suitable biassing configuration is shown for such a multi-electrode detector. The detector is disposed such that it is irradiated by the X-ray pulses on its cathode (114) side, and individual anode electrodes 112a—112f are disposed on the far side of CdTe wafer 110. Cathode 114 is connected to ground potential and each individual anode connected to a high voltage source through a respective resistor, suitably of the value 1 MΩ. The high voltage source is of the order of 50 volts. Each anode is AC coupled through a respective capacitor 26 to an associated amplifier/filter 20.

It should be appreciated that radiation can also be directed to the anode side of the detector. Further, a unitary anode and plural cathodes could be utilised to form the unitary array. This is equivalent to connecting electrode 114 to the high voltage source and grounding each of the electrodes 112a—112f.

Experiments show that crosstalk between the respective elements in the multi-element array of Figures 9B—9D are not appreciable. Figure 2 illustrates the effect of scanning a narrow (1.5 mm) beam over one of the individual electrodes of such a detector, having one electrode per 2 mm in the direction of scanning. It appears that CdTe exhibits a localised charge property in response to incident radiation. When taking beam width into account, scattering effects within the detector contributes to less than 1% crosstalk at 1 mm. Such crosstalk produces an effect analogous to smoothing the data over the 1 mm range. If desired, a simple digital de-convolution procedure can be utilised, in effect, to unfold the data, and thus eliminate the inter-element crosstalk effect.

Measurements of the electrical pulse output of the detectors as a function of detector bias was performed to ensure that all stopped photons contributed to the charge collected by the detector. It was found, as expected, that the response essentially flattens at high bias voltages where depletion is complete. Reference is made to Figures 13 and 14. It was also found that detectors with electrodes smaller than the detector area, in instances where the X-ray beam reaches edges, continue to yield larger pulses as the bias voltage is increased. This is apparently attributable to improved charge collection at the electrode edge. It was also found that where the incident radiation was collimated and only the central region of the electrode was irradiated, the detector output/detector bias response was the same as for the full coverage electrode detector.

It has also been determined that CdTe wagers only 2 mm thick provide adequate stopping power for CT scanning operations. The accompanying Table VI shows the stopping power of a 2 mm CdTe wafer in cooperation with various pre-hardening radiation filters.

TABLE VI

Stopping Power of 2 mm CdTe

150 kV X-Ray Beam

Prehardening Filter	Stopping Power
6 mm Al	96%
6 mm Al, 1.3 mm Cu	93%
6 mm Al, 2.6 mm Cu	90%
6 mm Al, 3.9 mm Cu	89%

Figure 15 shows a comparison of the stopping powers of 2 mm thickness of various detector materials used in CT scanning (including scintillator crystals). It should be noted that while bismuth germanate has a high stopping power, it is a scintillator as opposed to being a direct electrical conversion material.

It is sometimes desirable to measure radiation occurring at more than one effective energy level. Measurements at different energies can be taken by disposing two detectors in the path of the radiation, such that radiation passing through the first detector is detected by the second, etc. as illustrated in Figure 16. Such measurements can also be accomplished in accordance with one aspect of the present invention by disposing a multiple electrode detector, as shown in Figure 17 such that the radiation is incident on a surface of the detector transverse to the electrodes. Here, the first electrode develops a charge in accordance with the number of photons passing through a given distance of CdTe and the

second electrode develops a signal in accordance with the charge produced by the photons not blocked by the first portion of CdTe.

Figure 18 illustrates the usage of this invention in a CT scanner. A plurality of multi-element detector arrays 14 (Figures 11B—11D) are disposed in a stationary circular array. A pulsed radiation
5 source 24 provides a pulsed fan-shaped beam of radiation, which irradiates a body section 120 located   5
in a patient circle 122. Source 24, is made to rotate about body section 120 to effect a 180°—360° scan. The radiation passing through body section 120 is detected by detectors 14. Each element of detector 14 is AC coupled, through associated means 18, to an associated amplifier/filter 20. The respective amplifier/filters are multiplexed by a conventional multiplexer 124 through an analog-to-
10 digital converter 126 to suitable data processing circuitry 130. Data processing circuitry 130 may be   10
any suitably programmed data processor or computer for reconstructing a point-by-point electron density profile of the scanned body section from the individual measurements. It should be appreciated that since only one group of detectors is irradiated at a given time, a number of detectors (each from a different mutually exclusive group), can be interconnected to share a given amplifier/filter 20. The
15 scanning of the X-ray source itself effectively multiplexes respective interconnected detectors to a   15
common amplifier/filter 20.

In summary, it has been found that when used in accordance with the present invention, room temperature semiconductor detectors such as CdTe, can be advantageously used in CT scanning. This involves use of a pulsed radiation source and an amplifier/filter which provides a processed output pulse
20 from a predetermined range of frequency components present in the raw detector output. The range of   20
frequencies is chosen in accordance with the duration of the X-ray pulse. Such apparatus effectively removes the adverse properties of the detector as a factor in the overall performance. In effect, the pulse of incident radiation drives the rise-time response of the detector so that any adverse variations in detector rise-times are not apparent in the final output signal from the detector electronics interface.
25   Further, it appears that CdTe detectors in particular, are especially advantageous. Electrodes   25
appear to form only ohmic contacts with the CdTe wafer, without any appreciable boundary layer. The localised charge collection phenomenon also permits densely packed arrays of detectors to be formed on a single wafer by simple bath fabrication techniques. Densities of one detector per 2 millimeters have been defined on the face of a detector slab merely by scratching out sections of electrode material.
30   CdTe detectors are rugged, easily handled and, when used in accordance with the present   30
invention, very stable in electrical response. Since light pipes and PMT's are not needed, there is much more flexibility in designing detector configurations. For example, for dual energy detection, a thin CdTe detector (less than 0.5 mm thick) could be backed by a thicker, (2 mm) detector as shown in Figure 16. Alternatively, a unitary configuration could be utilised such as shown in Figure 17.
35   The same considerations that apply to CdTe also apply to H_gI_2. It is also thought that the following   35
semiconductor material among others, exhibit properties similar to those of CdTe and may be advantageously utilised in accordance with the present invention: gallium arsenide (GaAs), aluminium antimonide (AlSb), indium phosphate (InP), zince telluride (ZnTe), tungsten selenide, (WSe_2), cadmium selenide (CdSe), bismuth iodide (BI_3) and cesium antimonide (Cs_3Sb). Similarly, germanium (Ge) of low
40 quality typically exhibits property not unlike those of CdTe.   40

While various conductors shown interconnecting the elements of some of the figures of the drawing are shown in single line form, they are not shown in a limiting sense and may comprise plural connections as is understood in the art.

It should be further understood that the above description is of illustrative embodiments of the
45 present invention, and that the invention is not limited to the specific forms shown. For example, while   45
the present invention is described primarily in the environment of CT scanners, it should be appreciated that it may find application in other types of apparatus requiring the detection of radiation.

CLAIMS
1. A radiation detecting apparatus for use in detecting incident radiation pulses of predetermined
50 duration, said apparatus comprising a detector which generates electrical charge in response to incident   50
radiation, said detector being disposed to receive at least a portion of said radiation pulses, signal processing means AC coupled electrically to said detector for generating an output electrical signal accurately representing said incident radiation, said signal processing means including filter means for allowing only frequency components within a predetermined range of frequencies to contribute to said
55 output electrical signal, said range of frequencies being determined in accordance with the duration of   55
said incident radiation pulses so that said output electrical signal is undistorted by variations in the internal electrical characteristics of the detector.
2. Apparatus as claimed in claim 1 wherein the detector is a semiconductor detector.
3. Apparatus as claimed in either one of claims 1 or 2 wherein said signal processing means
60 comprises an electrical charge amplifier having an input electrically coupled to said detector through a   60
capacitor, a fast differentiator circuit connected in feedback relation with said charge amplifer, and an integrator responsive to the output of said charge amplifier.
4. Apparatus as claimed in claim 3 wherein said charge amplifier comprises a field effect transistor.

5. Apparatus as claimed in either one of claims 1 or 2 wherein said signal processing means comprises a charge amplifier electrically coupled to said detector through a capacitor; a differentiator responsive to the output signals of said charge amplifier; and an integrator responsive to the output of said differentiator.

5   6. Apparatus as claimed in either one of claims 1 or 2 wherein said signal processing means   5
comprises a charge amplifier electrically coupled to said detector through a capacitor; a differentiator responsive to the output signals of said charge amplifier; a voltage amplifier responsive to the output of said differentiator; and an integrator responsive to the voltage amplifier output.

7. Apparatus as claimed in any one of the preceding claims wherein said detector comprises a
10   wafer of a material selected from the group of cadmium telluride and mercuric iodide having a cathode   10
electrode in ohmic contact with one surface, and at least one anode contact in ohmic contact with an opposing surface of the wafer.

8. Apparatus as claimed in claim 7 further including means for developing a bias voltage across said wafer.

15   9. Apparatus as claimed in either one of claims 7 or 8 wherein said detector includes a plurality of   15
anode electrodes in ohmic contact with said wafer and electrically isolated from each other.

10. Apparatus as claimed in claim 9 wherein said detector is disposed with respect to said radiation source such that said radiation impinges upon a surface of said wafer transverse to said one surface, and said anode electrodes are respectively disposed such that they are at varying distances from said
20   radiation source, whereby each anode provides signals corresponding to differing energies of incident   20
radiation.

11. Apparatus as claimed in claim 2 further including a second semiconductor detector disposed to receive radiation passing through the first referenced semiconductor detector.

12. Apparatus as claimed in either one of claims 1 or 2 wherein said signal processing means
25   comprises a charge sensitive first stage having a first predetermined effective RC time constant, and a   25
second stage, cooperting with said first stage, having a second predetermined effective RC time constant; said first and second predetermined effective RC time constants being determined in accordance with the duration of said radiation pulses.

13. Apparatus as claimed in claim 12 wherein said first effective RC time constant is substantially
30   less than the duration of said radiation pulse and said second effective RC time constant is of the same   30
order as said pulse duration.

14. Apparatus as claimed in claim 12 wherein said first effective RC time constant is substantially greater than the duration of said radiation pulse and said second effective RC time constant is of the same order as said pulse duration.

35   15. Apparatus as claimed in any one of claims 1 to 5 wherein said detector means comprises a   35
wafer of a material selected from the group of cadmium telluride and mercuric iodide having a first electrode in contact with a first surface thereof, and at least one second electrode in contact with a second surface.

16. Apparatus as claimed in claim 15 wherein said detector means includes a plurality of said
40   second electrodes contacted to said water and electrically isolated from each other, whereby a plurality   40
of individual detectors are provided on said wafer.

17. Apparatus as claimed in claim 16 wherein said detector is disposed such that said radiation impinges upon a third surface of said wafer transverse to said first and second surfaces, and respective ones of said second electrodes are disposed at respectively varying distances from said radiation source
45   so that each second electrode provides signals at differing effective energies of incident radiation.   45

18. Apparatus as claimed in claim 16 wherein said detector means includes a second detector disposed to receive radiation passing through the first referenced detector.

19. Apparatus as claimed in any one of claims 16 to 18 wherein a plurality of said signal processing means are provided, each electrically coupled to a corresponding one of said separate
50   electrodes.   50

20. Apparatus as claimed in any one of claims 16 to 19 wherein the wafer thickness between said first surface and said second surface is approximately 2 mm.

21. Apparatus as claimed in any one of claims 16 to 20 wherein said electrically isolated electrodes are disposed on said second surface such that adjacent centres are relatively disposed at
55   approximately 1 mm intervals or less along the length of said second surface.   55

22. Apparatus as claimed in claim 21 wherein a plurality of said isolated electrodes are disposed along the width of said second surface.

23. Apparatus as claimed in either one of claims 1 or 2 wherein said signal processing means comprises a charge sensitive amplifier having a frequency response such that only frequencies within a
60   predetermined band of frequencies centred about a frequency approximately equal to the inverse of the   60
duration of said incident radiation pulses contribute to said output electrical signal.

24. Apparatus as claimed in either one of claims 1 or 2 wherein said signal processing means comprises a charge sensitive amplifier having a frequency response such that only frequencies within a predetermined band of frequencies centred about a frequency approximately equal to the inverse of
65   twice the duration of said incident radiation pulses contribute to said output electrical signal.   65

25. A radiation detector apparatus substantially as hereinbefore described with reference to Figures 4 to 7 and 9 to 18 of the accompanying drawings.

26. Apparatus for providing a measurement of radiation transmission/absorption by a cross-section of a body, said apparatus comprising a pulse radiation source operable to generate radiation pulses of predetermined duration, and a radiation detecting apparatus as claimed in any one of the preceding claims positioned to receive such radiation after passing through the body.

27. A tomographic scanner for providing a plurality of radiation beam transmission/absorption measurements taken along each of a plurality of paths with a cross-section of a three dimensional body and for computing from such measurements the relative radiation absorptions occurring at incremental areas disposed within said cross-section, said tomographic scanner comprising means for generating a pulsed beam of said radiation, said beam pulses being of predetermined duration; a radiation detecting apparatus as claimed in any one of claims 1 to 25; means for effecting a change in the geometric relationship between at least two of said radiation beam, detector, or body, such that said detector receives radiation from along a plurality of different paths through said body; and data processing means, receptive of said output signal for computing said relative radiation absorptions.

28. A tomographic scanner as claimed in claim 27 wherein a plurality of radiation detectors, each disposed to detect the radiation after passing through the body along a different path, are provided and wherein each detector has a respective signal processing means.

29. A method of detecting incident radiation pulses of predetermined duration, said method comprising the steps of disposing a semiconductor detector to receive at least a portion of said radiation, said semiconductor detector generating an electrical charge in response to said received radiation; generating an electrical output signal indicative of said charge; and filtering out components of said electrical output signal not in accordance with the duration of said radiation pulses so that said electrical output signal is substantially undistorted by the characteristics of said semiconductor detector.

30. A method for providing a measurement of radiation transmission/absorption by a cross-section of a body, comprising the steps of, generating pulses of radiation of predetermined duration; directing said radiation pulses to irradiate said body; disposing a semiconductor detector to receive at least a portion of said radiation transmitted through said body, said semiconductor detector generating an electrical charge in response to said received radiation; generating an electrical output signal indicative of said charge; and filtering out components of said electrical output signal not in accordance with the duration of said radiation pulses so that said electrical output signal is substantially undistorted by the characteristics of said semiconductor detector.

31. A method as claimed in claim 30 wherein a plurality of said semiconductor detectors are provided, said detectors comprising a wafer of material selected from the group of cadmium telluride and mercuric iodide disposed to receive at least a portion of said radiation transmitted through the body; one side of said wafer being formed with a plurality of spaced apart electrodes and on an opposing surface a common electrode, respective electrical charges being developed between said plurality of electrodes and said common electrode; and wherein said method further comprises generating respective electrical output signals indicative of said respective electrical charges; filtering said respective electrical output signals to develop respective filtered electrical output signals having only frequency components within a predetermined range of frequencies in accordance with the duration of said radiation pulses; and processing said filtered electrical output signals to provide an accurate representation of the intensity of the radiation.

32. A method as claimed in claim 31 for providing simultaneous measurements of radiation transmission/absorption in a plurality of adjacent cross-sections of a body, wherein each of said plurality of electrodes receives radiation transmitted through a respective one of said cross-sections, and develops a respective electrical charge in said wafer in accordance with said received radiation, so that signals are processed to provide respective representations of the transmission/absorption of said respective cross-sections.

33. A method as claimed in claim 31 for providing simultaneous measurements of radiation transmission/absorption of a body at a plurality of effective energy levels, wherein at least a portion of said radiation pulses transmitted through said body are incident on a surface of said wafer transverse to said one surface and said opposing surface to develop respective charges between said plurality of electrodes and said common electrode in accordance with the intensity of the radiation passing into the region of said wafer between respective ones of said plurality of electrodes and said common electrode, so that said respective filtered electrical output signals are processed to provide accurate representations of said radiation transmission/absorption at the energy levels associated with said radiation passing through respective distances of wafer material.

34. A method as claimed in claim 29 of detecting incident radiation pulses of predetermined duration substantially as hereinbefore described.

35. A method as claimed in claim 30 for providing a measurement of radiation transmission/absorption by a cross-section of a body, substantially as hereinbefore described.